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Effectiveness of Phantom Stimulation in Shifting the Pitch Percept in Cochlear Implant Users

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Objectives: Phantom electrode stimulation was developed for cochlear implant (CI) systems to provide a lower pitch percept by stimulating more apical regions of the cochlea, without inserting the electrode array deeper into the cochlea. Phantom stimulation involves simultaneously stimulating a primary and a compensating electrode with opposite polarity, thereby shifting the electrical field toward the apex and eliciting a lower pitch percept. The current study compared the effect sizes (in shifts of place of excitation) of multiple phantom configurations by matching the perceived pitch with phantom stimulation to that perceived with monopolar stimulation. Additionally, the effects of electrode location, type of electrode array, and stimulus level on the perceived pitch were investigated.

Design: Fifteen adult advanced bionics CI users participated in this study, which included four experiments to eventually measure the shifts in place of excitation with five different phantom configurations. The proportions of current delivered to the compensating electrode, expressed as σ, were 0.5, 0.6, 0.7, and 0.8 for the symmetrical biphasic pulses ($SBC_{0.5}$, $SBC_{0.6}$, $SBC_{0.7}$, and $SBC_{0.8}$) and 0.75 for the pseudomonophasic pulse shape ($PSA_{0.75}$). A pitch discrimination experiment was first completed to determine which basal and apical electrode contacts should be used for the subsequent experiments. An extensive loudness balancing experiment followed where both the threshold level (T-level) and most comfortable level (M-level) were determined to enable testing at multiple levels of the dynamic range. A pitch matching experiment was then performed to estimate the shift in place of excitation at the chosen electrode contacts. These rough shifts were then used in the subsequent experiment, where the shifts in place of excitation were determined more accurately.

Results: Reliable data were obtained from 20 electrode contacts. The average shifts were 0.39, 0.53, 0.64, 0.76, and 0.53 electrode contacts toward the apex for $SBC_{0.5}$, $SBC_{0.6}$, $SBC_{0.7}$, $SBC_{0.8}$, and $PSA_{0.75}$, respectively. When only the best configurations per electrode contact were included, the average shift in place of excitation was 0.92 electrode contacts (range: 0.25 to 2.0). While $PSA_{0.75}$ leads to equal results as the SBC configurations in the apex, it did not result in a significant shift at the base. The shift in place of excitation was significantly larger at the apex and with lateral wall electrode contacts. The stimulus level did not affect the shift.

Conclusions: Phantom stimulation results in significant shifts in place of excitation, especially at the apical part of the electrode array. The phantom configuration that leads to the largest shift in place of excitation differs between subjects. Therefore, the settings of the phantom electrode should be individualized so that the phantom stimulation is optimized for each CI user. The real added value to the sound quality needs to be established in a take-home trial.

Key words: Cochlear implant, Low-frequency, Phantom stimulation, Pitch, Pitch shift.

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INTRODUCTION

Cochlear implants (CIs) are electronic devices that partially restore hearing in severely hearing-impaired and deaf individuals. Although CI users can score up to 100% correct on speech tests in quiet (Rak et al. 2017), their understanding of speech in noisy environments declines drastically compared with normal hearing subjects. In addition, CI users report limited perceived sound quality and music appreciation (Kong et al. 2004). This is not surprising, as a lot of the frequency information from acoustic sound is lost during CI processing. For most of the speech coding strategies, the sound is filtered between approximately 70 and 7500 Hz; then, the envelope of the signal is extracted and delivered to the auditory nerve via 12 to 22 frequency bands, depending on the type of device. CI users often describe the perceived sound as very high pitched and sharp, which is a consequence of the fact that the electrode array typically is not inserted deep enough to stimulate the fibers physiologically tuned to the lower speech frequencies. It is well known that the transmission of low-frequency information, either electrically or acoustically, is important for speech perception and music appreciation (Hochmair et al. 2003; Qi et al. 2011; von Ilberg et al. 2011; Munjal et al. 2015). The current study investigated whether CIs place of stimulation can be shifted toward the apex when using phantom stimulation, which potentially can improve the transmission of low-frequency information.

In natural hearing, low-frequency sounds are coded in both place (place pitch) and time (rate pitch) in the apical region of the cochlea. It is well known that place pitch is coded across the complete CI electrode array, which is usually placed 1 to 1.5 turns in. While there is evidence that rate pitch is coded at both the base and the apex (Townshend et al. 1987; Carlyon et al. 2010), Landsberger et al. (2016) showed that low rates sound clean only at apical places of stimulation. CIs make use of the tonotopic organization of the cochlea (i.e., place pitch) by delivering low-frequency signals via the apically located electrode contacts and high-frequency signals via the basally located contacts. One way to transmit low-frequency signals is by stimulating the most apical regions of the cochlea. Therefore, CI electrode arrays are ideally inserted all the way up to the apex. This deep insertion, however, is limited because of the anatomy of the human cochlea, which becomes narrower toward the apex. In addition, other variables such as the characteristics of the electrode array itself (e.g.,

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stiffness, length, shape, and thickness), the experience of the surgeon, and anatomic abnormalities contribute to the relatively shallow insertion depth of most of the currently available electrode arrays (Cosetti & Waltzman 2012). For example, the HiFocus1J electrode array (Advanced Bionics, Valencia, USA) has a mean angular insertion depth of only 405 to 480° (van der Marel et al. 2014; Landsberger et al. 2015; van der Jagt et al. 2016). While almost full coverage of the cochlea is possible with the Med-El CI system (Innsbruck, Austria) (Buchman et al. 2014), there is conflicting evidence about the effect of this deep insertion. While Büchner et al. (2017) have shown a benefit of deeper inserted electrodes, Gani et al. (2007) and Kenway et al. (2015) showed that CI users hear better when the most apical electrodes are deactivated. Performance with long electrodes was better in Hochmair et al. (2003) when all 12 electrodes active or when a subset of electrodes was spanning the whole array (including the apex) compared with just electrodes 5 to 12 activated. Arnoldner et al. (2007) thoroughly investigated apical electrode deactivation in longer electrode arrays and conclude that having apical contacts is beneficial, but they should be more separated than they are now. However, other studies showed that the deeper electrode insertion could lead to trauma to the spiral ganglion cells (Wardrop et al. 2005; Finley et al. 2008; Kalkman et al. 2015), potentially limiting a beneficial effect.

As deep insertion of electrode arrays can be complex, many subjects are already implanted with relatively short electrode arrays, and it may be desirable to stimulate the more apical regions of the cochlea, alternative methods to deliver low-frequency information to the auditory nerve have been developed. Recent studies have shown that phantom stimulation can produce pitch percepts lower than that of the most apical electrode contact, without needing to further insert the electrode array (Saoji & Litvak 2010; Macherey et al. 2011; Macherey & Carlyon 2012; Klawitter et al. 2018). In phantom stimulation, two electrode contacts, one primary and one compensating, are simultaneously stimulated with opposite polarity. The current directed to the compensation electrode contact is a fraction (denoted as the current compensation coefficient σ) of that administered to the primary electrode contact (Fig. 1B); for example, $\sigma = 1$ means that the amplitude at the compensating contact is equal to the amplitude at the primary contact (i.e., bipolar stimulation), $\sigma = 0.5$ means that the amplitude is 50% of that at the primary contact, and $\sigma = 0$ equals monopolar (MP) stimulation. The center of the electrical field, and therefore the perceived pitch, is steered toward the apex because of electrical field shaping with phantom stimulation (Saoji et al. 2013). Multiple studies demonstrated significant pitch shifts toward the apex with phantom stimulation that used symmetric biphasic (SB) pulse shapes. For example, Saoji and Litvak (2010) found shifts of 0.5 to 2.0 electrode contacts toward the apex when using the σ values that led to the greatest shift in place of excitation for each subject. These optimal σ values varied greatly from 0.38 to 0.88, implying that, in some subjects, the shift is smaller at higher σ values. Moreover, one of the subjects heard two distinct pitches at $\sigma = 1$, which equals bipolar stimulation. It was hypothesized that, at these higher σ values, the compensating electrode contact also generates a secondary peak that causes excitation of fibers near the compensating electrode (Saoji & Litvak 2010; Macherey & Carlyon 2012). The extra peak, or side lobe, can counteract the effect of lowering the pitch, and could also explain the reported dual pitched tone. Simulations with our 3D computer model of the human cochlea (Snel-Bongers 2013) also point in this direction.

To avoid this side-lobe phenomenon, pseudomonophasic (PS) pulses were used (Macherey et al. 2011). These pulses consist of a short- and high-amplitude phase, followed by a long- and low-amplitude phase (Fig. 1C). When the pulse of the primary electrode contact starts with an anodic phase, it is expected that this contact excites the auditory nerve, while the compensating contact does not. This hypothesis is based upon the finding that anodic current and high amplitudes are more effective in exciting the human auditory nerve than cathodic pulses (Macherey et al. 2008). In line with this hypothesis, Macherey et al. (2011) presented stimuli in narrow bipolar mode and showed that place-pitch was lower when the more apical electrode was stimulated with the anodic-first PS pulse shape than with the cathode-first PS pulse shape.

Although previous studies showed that phantom stimulation can result in a shift in place of excitation toward the apex, the size of the effect is unknown. Saoji and Litvak (2010) compared multiple SB stimulation modes in pitch-ranking experiments to find the configuration that led to the largest shift for each individual subject. The shift in excitation resulting from this best configuration was quantified with a two-interval, forced-choice procedure. However, because only the best configurations were examined, their results do not include the average pitch in place of excitation per phantom configuration. Macherey et al. (2011) and Macherey and Carlyon (2012) compared both SB and PS configurations at multiple pulse rates in pitch-ranking experiments. Such experiments provide information about the direction of the shift relative to the other configurations and which of the tested configurations lead to the largest shift, but only limited information about the size of the shift in place of excitation. Macherey et al. (2011) did compare the pitch perceived with PS and MP stimuli to that perceived on the contralateral normal-hearing ear of two participants and found the pitches to decrease from 1077 to 976 Hz and from 988 to 811 Hz when using the PS stimulation mode. Unfortunately, the low number of subjects limited the interpretation of the results. Therefore, the current study quantified the shift in place of excitation after phantom stimulation with multiple configurations by pitch-matching the phantom stimuli to MP (or current steered) stimuli in two-alternative forced choice tasks in 15 subjects. To improve the reliability of the pitch matching experiments, only the electrode contacts for which the CI users had a relatively high pitch discrimination were tested. A necessary limitation the method used here and elsewhere (Saoji & Litvak 2010) is that we cannot measure pitch shifts beyond the end of the electrode array, and so instead present phantom stimuli to pairs of electrodes which are close to, but not at the apical end of the array. We assume that the pattern of pitch shifts observed would generalize to stimulation involving the most apical electrode.

When incorporating phantom stimulation in a speech coding strategy, shifts in place of excitation due to variations in intensity could interfere with the perceptual outcome; thus, the effect of stimulus level must be studied. While previous studies about phantom stimulation focused on the effect of different configurations, to the best of our knowledge, no data are available about the effect of stimulus level on the perceived pitch. Previous studies report contradictory results about the effect of the stimulus level on pitch perception for MP stimuli. Arnoldner et al. (2006) and Carlyon et al. (2010) demonstrated an increased pitch with increasing stimulus level, while others found a significant

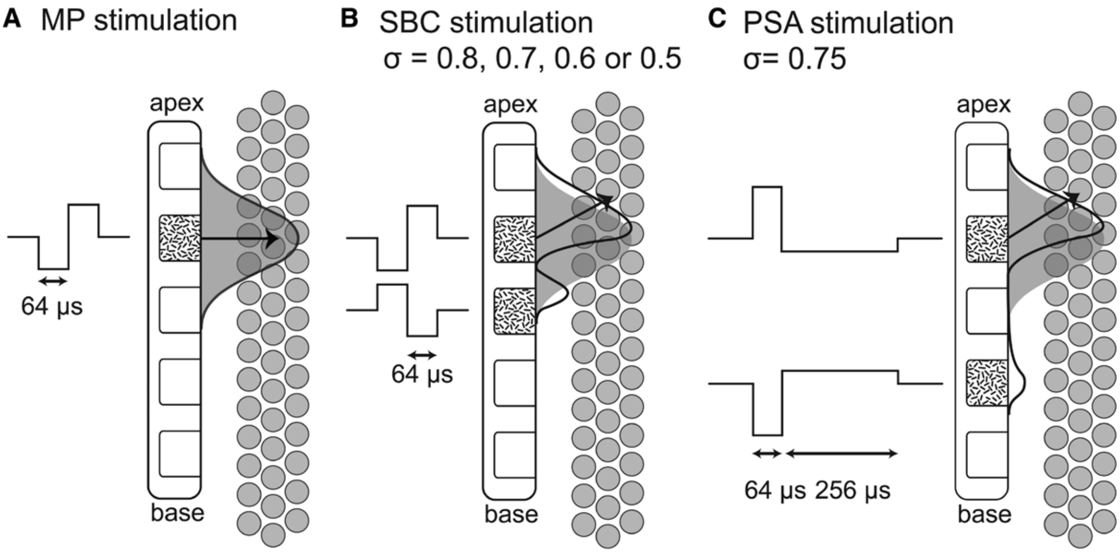

Fig. 1. Schematic illustration of different stimulation techniques. Spiral ganglion cells are illustrated as circles. The activated electrode contacts are filled, and the (A) electrical field from monopolar (MP) stimulation is shaded gray. The shape of the expected electrical field of the phantom (PE) symmetric biphasic cathodic first pulses (SBC) (B) and pseudomonophasic anodic first pulse (PSA) (C) are displayed as black lines. The direction of the expected shift in place of excitation with the different stimulation modes is indicated by the arrow.

decrease in pitch when the MP stimulus level increased (Townshend et al. 1987; Pijl 1997; Kalkman et al. 2014). The shift in place of excitation following phantom stimulation was modeled as a function of stimulus intensity in our computational model of the human cochlea (Snel-Bongers 2013). The model predicted that the pitch shift relative to MP stimulation was dependent on the stimulation level indeed, with higher stimulation levels leading to a larger shift toward the apex. To study this, the pitch-matching experiments were repeated at multiple stimulus levels. The current study compared the effect size (in shift in place of excitation) of multiple phantom configurations by matching the perceived pitch to that from MP stimulation. Additionally, the effects of the electrode location, the type of electrode array, and the stimulus level on the perceived pitches were investigated.

MATERIALS AND METHODS

Subjects

All 15 subjects were unilaterally implanted with an Advanced Bionics HiRes90K or Clarion CII device, with either the HiFocus 1J, HiFocus Mid-scala, or HiFocus 1J with positioner electrode array (Sylmar, CA). Only postlingually deaf CI recipients were included in this study. The mean phoneme score on open-set Dutch monosyllabic (CVC) word tests taken in quiet conditions at 65 dB was 88.7% (range: 78 to 98%). All subjects had all 16 active electrode contacts in their clinical strategy. Table 1 displays the characteristics of each subject. The study protocol was approved by the committee for Medical Ethics of the Leiden University Medical Center (P02.106 AC).

Phantom Stimulation

The concept of both MP and phantom stimulation is schematically illustrated in Figure 1. In the current study, the apical electrode contact was designated as the primary electrode and the basal electrode as the compensating one. The SB pulse shape had a cathodic phase first on the primary electrode (therefore denoted as SBC, Fig. 1B), and the primary and compensating electrode contacts were adjacent to each other. The PS pulse shape was asymmetrical, so that the amplitude of the first phase (which was anodic, therefore denoted as PSA) was four times as high as the second phase, and the duration of the second phase was four times longer than the first (Fig. 1C). The amplitude of the second PSA phase was reduced fourfold to maintain charge balancing. The two stimulated electrodes were spaced one electrode from one another, identical to the configuration used in Macherey and Carlyon (2012). The compensation coefficient (*σ*) values used in this study are depicted in Table 2 and were chosen because these configurations were assumed to result in the largest shift in place of excitation without causing pitch reversal (Saoji & Litvak 2010; Macherey et al. 2011). The default pulse durations were 64 µs for the first phase (in both SBC and PSA modes), 64 µs for the second phase in SBC, and 256 µs in PSA mode. The stimulus level was measured in clinical units (CUs), according to the formula CU = pulse duration (µs) · amplitude (µA)/78.7. If the stimulus level exceeded the compliance level of 7.2 Volt, the phase duration was increased by 10.8 µs per phase to increase the charge (and by 43.2 µs for the second phase of the PSA mode). Stimuli were 300 ms long, with pulse rates of 1400 pulses/s, and the time between stimuli was 500 ms.

Experiments

Four experiments were conducted using a custom-made MATLAB (Mathworks, Inc., Natick, MA) interface and the Advanced Bionics' research tool BEDCS (Bionic Ear Data Collection System, Advanced Bionics, Sylmar, CA). (1) A pitch discrimination experiment was first completed to determine which basal and apical electrode contacts should be used for the subsequent experiments. (2) An extensive loudness balancing experiment followed to determine the threshold level (T-level) and most comfortable level (M-level) and enable testing at multiple well-defined levels of the dynamic range. (3) A pitch-matching experiment was then performed to estimate the shift in place of excitation at the chosen electrode contacts. (4) These rough shifts were then used in the final experiment to more accurately determine the shifts in place of excitation.

TABLE 1. Characteristics of the study subjects

Subject	Gender	Age	CVC (Ph%)	Etiology	Duration of Deafness (yrs)	CI Experience (yrs)	Implant	CI Side	Tested Electrode Contacts (Insertion Angle in Degrees)	Pulse Width (µs)
S03	Male	59	78	Unknown	4	2	HiRes 90K HiFocus MS	R	4 (300), 12 (123)	64.6
S06	Male	59	93	Hereditary	30	16	Clarion CII HiFocus I with positioner	R	4 (-), 12 (-)	64.6
S07	Female	66	87	Unknown	48	10	HiRes 90K HiFocus 1J	R	12 (124)	64.6
S08	Male	72	91	Otosclerosis	23	9	HiRes 90K HiFocus 1J	R	5 (348), 13 (137)	64.6
S09	Female	68	94	Unknown	24	16	Clarion CII HiFocus I with positioner	L	11 (-)	64.6
S10	Male	70	80	Meniere	21	4	HiRes 90K HiFocus 1J	R	12 (186)	75,4
S11	Female	71	83	Hereditary	18	3	HiRes 90K HiFocus 1J	R	6 (303)	97.0
S12	Female	74	88	Unknown	13	8	HiRes 90K HiFocus 1J	R	9 (209)	64.6
S13	Female	49	83	Meningitis	1	14	Clarion CII HiFocus I with positioner	R	7 (275)	64.6
S14	Female	64	91	Hereditary	8	7	HiRes 90K HiFocus 1J	L	5 (428), 13 (191)	86.2
S15	Female	55	98	Unknown	19	16	Clarion CII HiFocus I with positioner	L	7 (-), 10 (-)	64.6
S16	Female	64	96	Unknown	15	16	Clarion CII HiFocus I with positioner	L	5 (-), 12 (-)	86.2
S17	Male	62	92	Noise-Induced	4	13	HiRes 90K HiFocus 1J	R	5 (-)	75.4
S19	Male	66	90	Hereditary	2	3	HiRes 90K HiFocus MS	R	7 (180), 14 (61)	64.6
S20	Female	66	86	Hereditary	14	2	HiRes 90K HiFocus MS	L	8 (153)	64.6

CVC, consonant-vowel-consonant; L, left ear;Ph%, percentage phonemes correct on a standard monosyllabic (CVC) word test at 65 dB; R, right ear.

Pitch Discrimination Experiment • In this three-alternative forced-choice task, adapted from Biesheuvel et al. (2019), subjects were asked to differentiate a target stimulus from two identical reference stimuli without receiving feedback about the correct answer. Both the target and reference stimuli consisted of 300-ms pulse trains with biphasic pulses. Pulse durations were 32 µs per phase and pulse rates were 1400 pulses/s. All stimuli were presented at the M-level, which was determined using the eight-point loudness scale described by Potts et al. (2007). If the M-level (in CU) exceeded the compliance limit, the pulse duration was increased in increments of 10.78 µs. The M-level on each electrode contact was loudness balanced with the apically adjacent electrode contact. To avoid confounding effects from potential loudness cues, a level roving of ±10% relative to M-level was applied to each stimulus. The target stimulus was based on the current steering, which involves the simultaneous stimulation of two adjacent electrode contacts, thereby creating an intermediate pitch percept (Firszt et al. 2007). Snel-Bongers et al. (2012, 2013) showed that current steered and MP are equivalent with regard to spread of excitation, channel interaction, and threshold levels. The proportion of the total current directed to the basal contact was denoted as α, and the proportion to the apical contact as $1 - \alpha$. The target stimulus had α values ranging from 0.25 to 1, whereas α for the reference stimuli was 0 (apical electrode only). Initially, the experiment was repeated five times for each electrode pair, with the target stimulus having an $\alpha = 1$ (basal electrode only), that is, the spatial difference between the target and reference stimuli was one electrode contact. If the percentage correct exceeded 66% for a certain electrode pair, the test was repeated at a more difficult ratio: with the distance between the target and reference stimuli halved ($\alpha = 0.5$). If the score at this ratio was still >66%, $\alpha = 0.25$ was tested. The final pitch discrimination score for each electrode pair was calculated as follows:

$$\text{Pitch discrimination score} = K - (\text{proportion correct} \cdot L)$$

with K being the lowest α at which the score was ≥66.6% (which was 2.0 if the score was <66.6% at 1.0) and L being the lowest measured α, which was always half of that corresponding to K. The proportion correct refers to the score with the lowest measured α, which was never higher than 1. For example, if one scored 3/5 correct at $\alpha = 0.25$, the pitch discrimination score = $0.5 - (3/5 \cdot 0.25) = 0.35$. The best possible score was 0.25 and the worst possible score 2.0. The shift in place of excitation caused by phantom stimulation was expected to be approximately one electrode contact apical to the main electrode contact (Saoji & Litvak 2010; Carlyon et al. 2014; Klawitter et al. 2018). Therefore, we first identified the best-performing pairs of electrodes (the lowest α), and as a default selected the electrode immediately basal to this pair for further testing. However, if this was electrode 3 or lower we chose to continue with electrode 4, for example, in S3. Moreover, if a complete region had α-scores below 0.5, the middle electrode contact was chosen to obtain the most reliable results (e.g., in S16). Only electrode pairs with an $\alpha \leq 1.0$ were used for further

TABLE 2. Configurations used in this study

Pulse Shape	Compensation Coefficient σ	Denotation
Symmetrically biphasic	0.5, 0.6, 0.7, 0.8	$SBC_{0.5}$, $SBC_{0.6}$, $SBC_{0.7}$, and $SBC_{0.8}$
Pseudomonophasic	0.75	$PSA_{0.75}$

PSA, pseudomonophasic anodic first pulse shape; SBC, symmetrically biphasic cathodic first pulse shape.

testing to ensure that the subjects were capable of undergoing subsequent testing.

Loudness Balancing Experiment • To estimate the dynamic range, both threshold levels (T-levels) and M-levels were determined for all electrode contacts, in all the configurations depicted in Table 2 and in the MP mode. First, the impedances of all 16 electrode contacts were measured to determine the voltage compliance limit. At higher σ values, the pulse duration, rather than the current level, may need to be increased to reach equal loudness within the compliance limits of the device (Saoji & Litvak 2010). To keep pulse durations equal across all configurations, the highest σ level ($SBC_{\sigma=0.8}$) pulse duration was set as the standard pulse duration for all experiments for each subject (see Table 1). The T- and M-levels were determined using the same eight-point loudness scale used in the pitch discrimination experiment (Potts et al. 2007; Biesheuvel et al. 2019). Two ascending (starting at loudness level 1) and two descending (starting at loudness level 7) trials per electrode contact were performed and the average M-levels were calculated. The SBC and PSA M-levels were balanced with the MP M-levels by sequentially presenting the two stimuli, and adjusting the MP current level until equal loudness was achieved.

Pitch-matching Experiment • The pitch matching experiment, which used a 2-up-2-down procedure, was conducted to estimate the shift in place of excitation for all the SBC and PSA configurations at two locations along the electrode array, one basal and one apical electrode contact determined in the pitch discrimination experiment. Two stimuli (one (steered) MP and one phantom stimulus, in random order) were administered and the subject was asked if the second stimulus sounded lower, higher or equal in pitch compared to the first one, without receiving feedback about it. At the initiation of the experiment, the (steered) MP stimulus was delivered three electrode contacts apical or one electrode contact basal to the main electrode contact, in random order, while the location of the phantom stimulus remained constant. The place of stimulation of the steered MP stimulus was gradually shifted toward the main electrode contact (apically or basally depending on the starting point) in step sizes of 0.05 α using current steering, until the two stimuli were perceived as equal in pitch. Then, the place of stimulation of the steered MP signal was shifted beyond the main contact until it produced a pitch distinctive from the reference. Next, the steered MP stimulus was shifted back to where the MP and phantom signal were perceived equal in pitch again, after which the experiment was terminated. The stimulus level (calculated in µA) was roved by 10% to prevent loudness cues from influencing the results. The experiment was repeated four times per configuration, and the average was used as the reference electrode Contact in the final pitch shift experiment.

Pitch Shift experiment • In this two-alternative forced-choice task, based on experiment 4 in the article by Saoji and Litvak (2010), the pitch percept of a phantom stimulus was compared with that of current steered MP stimuli. The MP stimuli were presented at the (virtual) electrode contacts that were 0.0, 0.25, 0.5, 0.75, 1.0, 2.0, and 4.0 contacts from the reference electrode contact in both directions (apical and basal), determined in the pitch matching experiment. The phantom and MP stimuli were sequentially presented in a random order, after which the subjects were asked to indicate which of the two stimuli was higher pitched, with no feedback about the right answer provided. Each phantom configuration was compared with the 13 MP stimuli and repeated 15 times, resulting in blocks of 195 trials. The configurations were tested in a random order, to exclude the role of fatigue or learning. The stimuli were presented at the M-level, determined in the loudness balancing experiment, with loudness roving of 10%. To test the effect of stimulus level on shift in place of excitation, the experiment was repeated at 75 and 50% of the M-level in CUs. Due to time constraints, the effect of stimulus level was only evaluated with the PSA configuration and the SBC configuration that resulted in the largest shift in place of excitation for that specific subject.

Data Analysis

To quantify the degree of shift in place of excitation, each subject's data were fit with a cumulative Gaussian psychometric function using the "psignifit" algorithm (Wichmann & Hill 2001a,b). To assure only true shifts were measured, the reliability of the psychometric functions was assessed. It was assumed that subjects are able to discriminate stimuli that are spatially separated two electrode contacts from each other, and this was confirmed in the pitch discrimination experiment. Therefore, subjects should be able to correctly indicate if the MP stimulus was higher pitched than the phantom stimulus, or not when the steered MP stimuli were two or four electrodes separated from the reference electrode contact. If the subject was unable to reach a correct score of at least 66.6% in these comparisons, the measurement was considered unreliable and was discarded from the analysis. This was the case for three phantom configurations for (subject number-electrode number) S07-E12, S11-E06, and S12-E09, two configurations for S19-E7, and one configuration for S03-E04, S06-E04, S06-E12, and S16-E12. All data for S10-E12 and S19-E14 were excluded because 6/9 and 5/9 measurements, respectively, did not meet the reliability rules. Ten out of the 15 discarded measurements were obtained at lower stimulus levels (50% or 75% of the M-level). This was in line with our expectations, as the task difficulty increases at lower stimulus levels. In total, data were obtained for 14 subjects, six of whom were measured at two locations along the electrode array. This means that data were obtained for 20 electrode contacts. There were nine measurements per electrode contact (five at the most comfortable loudness level and 2 at the two other levels), resulting in a total of 180 measurements, of which 15 were discarded because of the reliability rules described above.

Statistical Analysis

The *psignifit* software package for MATLAB provides a shift in place of excitation value with a confidence interval per tested setting. To account for the uncertainty in the measurement process and the reliability of each measurement, multiple imputations were made. Measurements were imputed independently from the normal distribution corresponding to the confidence interval. All further analyses were performed on the 10 imputed data sets and final results were based on pooling, using Rubin's rule implemented in SPSS (Harel & Zhou 2007; Van Buuren 2018). This process leads to the attenuation of any potential significance as imputed values reflect the measurement error involved in the shifts and thereby increases robustness of the analyses. For the pitch matching experiment, a linear mixed model analysis was used. A linear mixed model takes into account that measurements taken on an individual are more similar than measurements taken on different individuals. Furthermore, it corrects for missing data. Pitch shift values were checked for normality using histograms and did not show

TABLE 3. Pitch discrimination scores per electrode pair

Subject	Electrode Pair															Selected Primary Electrodes
	1–2	2–3	3–4	4–5	5–6	6–7	7–8	8–9	9–10	10–11	11–12	12–13	13–14	14–15	15–16	
S3	1.4	0.4	0.8	1.4					1.4	1.4	0.7	2				4, 12
S6	0.8	0.35	0.8	0.35					0.35	0.4	0.3	0.7				4, 12
S7	1.6	1.8	1.6	1.8	1.4	1.6	1.6	0.9	1.4	0.8	0.8	1.4	0.7	1.4	0.45	12
S8	1.6	0.8	0.8	0.45	0.7	0.8	0.4	0.7	0.7	1.4	0.4	0.45	0.35	0.7	1.4	5, 13
S9	1.4	2	0.8	1.4	1.6	0.8	1.6	1	0.8	0.45	1.4	1.4	0.7	1.8	0.8	11
S10	1.6	1.4	1.4	1.8	1.4	0.7	1.4	0.8	1.4	1	0.4	0.9	1.4	0.7	0.4	12
S11	0.8	1.6	1.4	1.4	0.8	0.8	1.6	1.4	0.8	1.4	1.6	1.8	1.4	1.4	0.8	6
S12	1.6	1.4	0.8	0.8	1.4	1.4	0.9	0.4	1.4	0.9	0.9	1.4	1.4	0.9	1.4	9
S13	0.8	1.6	1.4	1.6	1.8	0.4	1.6	1.8	1.6	0.8	1.8	1.6	1.6	1.6	1.6	7
S14	0.35	0.35	0.45	0.35	0.4	0.4	0.45	0.35	0.7	0.8	0.4	0.3	0.35	0.7	0.8	5, 13
S15	0.8	1.8	1.4	1.4	0.7	0.4	1.4	0.9	0.4	1.6	0.7	1.4	1.8	1.8	0.7	7, 10
S16	1.6	0.8	0.4	0.3	0.35	0.4	0.7	0.7	0.3	0.4	0.4	0.3	0.45	0.7	0.8	5, 12
S17	1.6	1.4	1.4	0.8	1.4	1.4	1.6	2	1.4	1.4	1.6	1.6	1.8	1.6	0.9	5
S19	1.6	0.7	1.4	1.6	0.8	0.7	0.9	1.4	0.9	1.6	1.4	0.8	0.8	1.4	1.8	7,14
S20	0.8	0.45	1.4	1.4	1.4	0.7	0.5	0.9	0.9	2	1.4	1.8	1.4	0.7	2	8

The electrode pairs around the electrode contacts that were selected for further assessment are shaded gray. Missing data are shaded light gray.

deviations. All data were analyzed with SPSS 23 (Statistical Package for the Social Sciences, SPSS Inc., Chicago, IL) software.

RESULTS

Pitch Discrimination

The results of the pitch discrimination experiment are shown in Table 3. For example, S7 had scores >1.0 on many electrode contacts, meaning that this subject could only discriminate two stimuli when the stimulation sites were separated more than one electrode from each other. The gray shaded electrode contacts were those that were located basally from the best functioning electrode contacts and were chosen for subsequent testing. Only electrode contacts with a score of 1.0 or less, or that had at least one adjacent electrode contact with a maximum $\alpha = 1.0$ were selected for subsequent testing. The far right column in Table 3 lists the selected electrodes. Due to time constraints, E5-E9 and E13-E16 were not tested for subjects 3 and 6.

Reliability of the Experiments

The individual pitch matching results are combined with the individual results of the pitch shift experiment in Figure 2. The outcomes of the pitch shift experiment are shown by the dots plus error bars, and the pitch-matching experiment outcomes are demonstrated by the crosses. The pitch-matching reference point usually fell within the error bars of the pitch shift experiment when there was SBC stimulation, showing that the pitch matching experiment had an additional value for the reliability of the pitch shift experiment. With PSA stimulation, the results of the two methods of measuring the shift in place of excitation differed more often, although this difference was not statistically significant when compared with any of the SBC configurations. The reference pitch was lower than the results from the pitch shift experiment during six of the 11 apical measurements.

Effect of Phantom Configuration on the Mean Shift in Place of Excitation

Figure 3 depicts the mean shift in place of excitation per phantom configuration and electrode location at the M-level, calculated with a linear mixed model analysis. Subjects confirmed that all of the presented stimuli were perceived as a single sound with a reasonably clear pitch. The mean shifts in place of excitation were 0.39 (SE = 0.14), 0.53 (SE = 0.14), 0.64 (SE = 0.15), and 0.76 (SE=0.14) electrode contacts toward the apex for $\text{SBC}_{0.5}$, $\text{SBC}_{0.6}$, $\text{SBC}_{0.7}$, and $\text{SBC}_{0.8}$, respectively; that is, the shift increased with increasing σ value for the SBC configurations. For $\text{PSA}_{0.75}$, the mean shift was 0.53 (SE = 0.14) electrode contacts. A linear mixed model with electrode location and phantom configuration as factors showed that only the shifts of $\text{SBC}_{0.5}$ and $\text{SBC}_{0.8}$ were statistically significantly different from one another ($p = 0.005$). Shifts in place of excitation with all phantom configurations were statistically significantly different from 0 (MP stimulation) ($p < 0.01$ for all configurations).

The largest shifts in place of excitation per electrode contact were most often obtained with $\text{SBC}_{0.7}$ and $\text{SBC}_{0.8}$ (both six times), followed by $\text{SBC}_{0.6}$ and $\text{PSA}_{0.75}$ (both three times), and least often with $\text{SBC}_{0.5}$ (two times). When only the results with the largest shifts per electrode contact were included in the analysis, as in Saoji and Litvak (2010), the mean shift was 0.92 electrode contacts toward the apex, with a minimum of 0.25 and a maximum of 2.0 electrode contacts. Because this data selection exaggerates the effect on pitch shift, all other analyses were performed on the data with all σs included. The mean shift in place of excitation at the apex was significantly larger (0.66, SE = 0.12) than at the basal part of the electrode array (0.47, SE = 0.12) ($p = 0.04$) (Fig. 3B). Therefore, the effect of configuration was analyzed separately for the apical and basal electrode contacts (Fig. 3C and D). A linear mixed model revealed that the different phantom configurations resulted in different shifts in place of excitation when phantom stimulation is applied to the more basally located electrode contacts. Both $\text{SBC}_{0.5}$ and $\text{SBC}_{0.8}$ ($p = 0.005$) and $\text{SBC}_{0.8}$ and $\text{PSA}_{0.75}$ ($p = 0.001$) differed significantly at the base, while no significant differences were found at the apex. Interestingly, $\text{PSA}_{0.75}$ stimulations did not result in a significant shift in place of excitation at the base ($p = 0.47$) but led to a shift of 0.8 electrode contacts ($p < 0.001$) at the apical electrode contacts. Moreover, the pitch shift with $\text{PSA}_{0.75}$ differed significantly between the two electrode locations

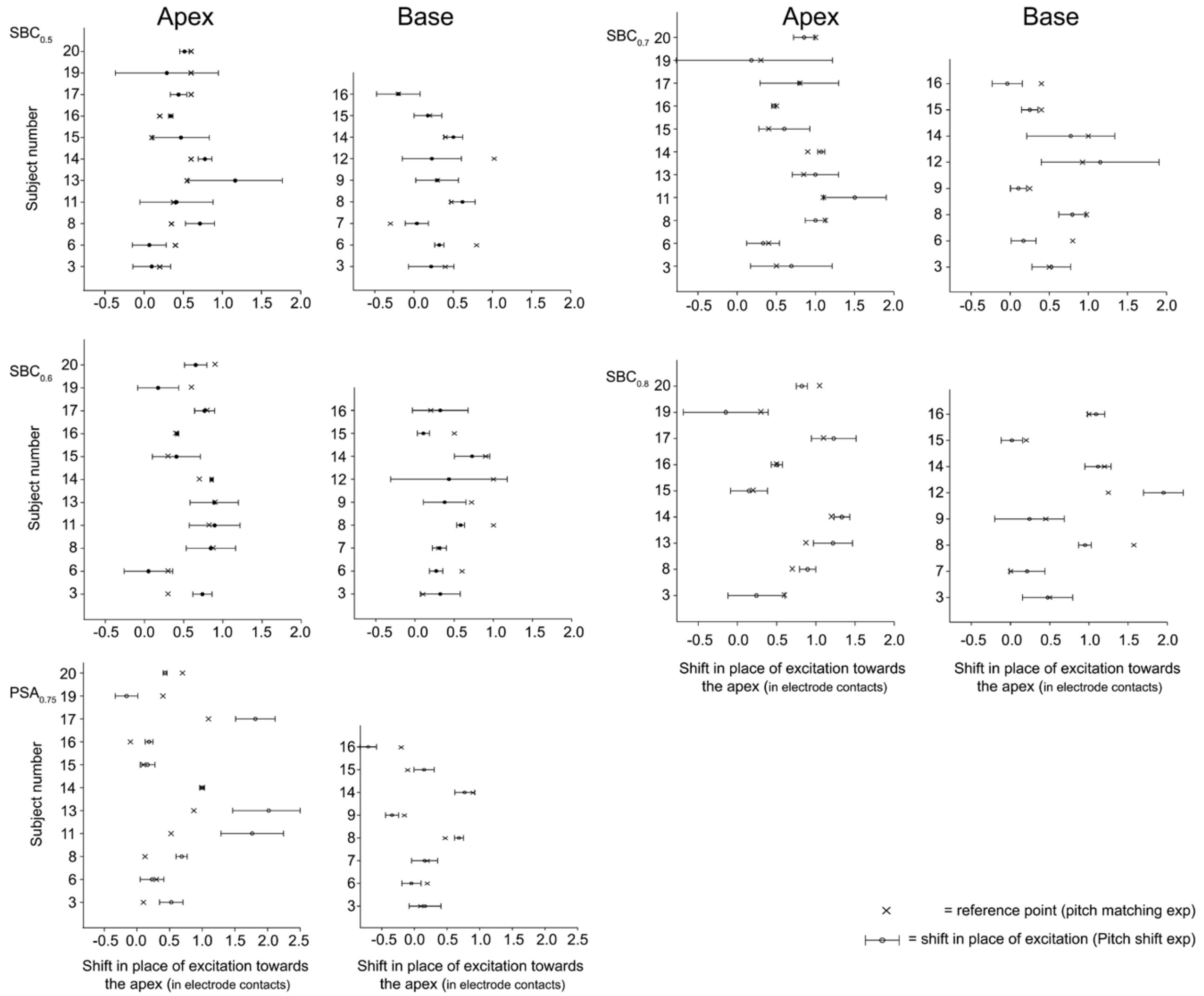

Fig. 2. Individual pitch matching and pitch shift experiment results for the SBC and the PSA configurations at the most comfortable level. The X is the reference point measured in the pitch-matching experiment. The circles with error bars show the mean and standard errors of the shift in place of excitation measured in the pitch shift experiment. PSA indicates pseudomonophasic pulse shape, anodic first; SBC, biphasic cathodic first pulse shape.

($p = 0.017$). Also, no significant shift was achieved at the base ($p = 0.13$) with $\text{SBC}_{0.5}$. All other configurations were statistically significantly different from steered MP stimulation at both the apical and basal electrode contacts.

As the electrode location influences the shift in place of excitation and the electrode type is known to influence the location of the electrode contacts, an additional analysis of the effect of electrode type on the shift was performed. Electrode types were divided into (semi) medial wall (HiFocus 1 with positioner or HiFocus MS, N = 8) and lateral wall (N = 7) electrodes. Because the HiFocus 1 with positioner electrode array is positioned closer to the modiolus than without positioner (van der Beek et al. 2005), subjects implanted with this electrode array were assigned to the (semi) medial wall group. We are aware of the fact that the HiFocus 1 with positioner and the HiFocus MS differ from each other in multiple ways, but in terms of distance to the inner wall they are rather similar. Because of the limited study group size, we clustered these two electrode types. The results are displayed in Figure 4. The shifts for the lateral wall electrode arrays (0.83 electrodes towards the apex) were significantly larger than the shifts for the medial wall electrode arrays (0.39 electrodes towards the apex, $p = 0.029$). Moreover, the lateral wall electrode arrays significantly differed from steered MP stimulation ($p < 0.001$), while the (semi) medial wall arrays did not ($p = 0.12$). When comparing the two electrode types separately, the configuration type did not influence the shift in (semi) medial wall electrode arrays, while higher degrees of phantom stimulation (higher σ values) did lead to larger shifts in lateral wall electrodes. Specifically, $\text{SBC}_{0.8}$ (1.16, SE = 0.19) led to a significantly larger shift in place of excitation than both $\text{SBC}_{0.5}$ (0.46, SE = 0.19, $p < 0.001$) and $\text{SBC}_{0.6}$ (0.68, SE = 0.19, $p = 0.02$). Also $\text{PSA}_{0.75}$ (1.02, SE = 0.2) showed a significantly larger shift ($p = 0.005$) than $\text{SBC}_{0.5}$. The mean shifts in place of excitation at three stimulus levels are displayed in Figure 5 for both the SBC and PSA configurations. For SBC stimulation only, the configuration that led to the largest shift was tested at stimulus levels other than the M-level. A linear mixed model with the factors “stimulus level” and “configuration” revealed no significant differences between the shifts at the M-level versus 75 and 50% of the M-level.

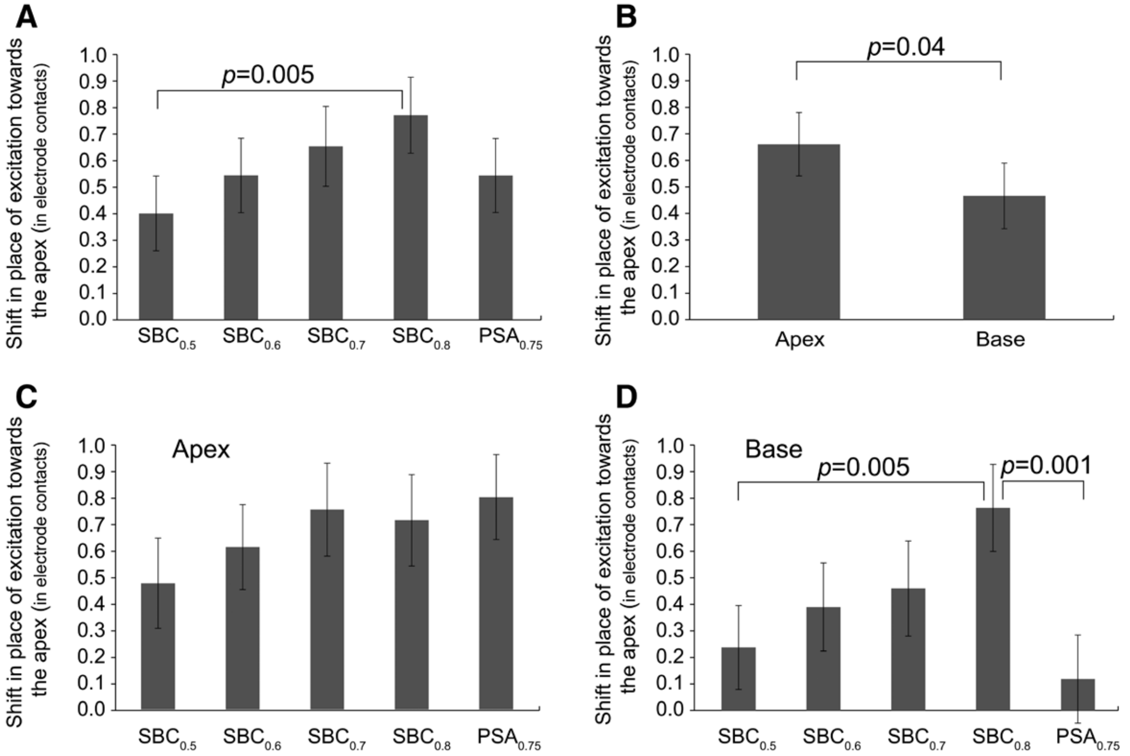

Fig. 3. Mean pitch shift, in electrode contacts, from the main electrode contact towards the apex at the M-level. The effects of different phantom configurations (A), electrode location (B), and different phantom configurations per electrode location (C and D) on the mean shift in place of excitation. All mean and standard errors were calculated using a linear mixed model analysis with either "configuration," "electrode location," or "configuration and electrode location" as factors. The error bars represent the standard error. The displayed means were compared in a more extensive linear mixed model, as described in the Results section, and only statistically significant *p* values (<0.05) are displayed. PSA indicates pseudomonophasic pulse shape, anodic first; SBC, biphasic cathodic first pulse shape.

DISCUSSION

This study confirmed that both the SBC and PSA phantom stimulation modes cause a statistically significant shift of the pitch percept toward the apex. Of the tested phantom configurations, $SBC_{0.8}$ caused the largest average shift in place of excitation of 0.76 (SE = 0.14) electrode contacts. However, there was a great variability within and between subjects. A higher degree of phantom stimulation (i.e., higher σ value) does not always cause a larger shift in place of excitation, which is consistent with previous studies that describe pitch reversals with monotonically increasing σ following phantom stimulation (Saoji & Litvak 2010; Macherey & Carlyon 2012). When only the configurations that resulted in the largest shift for each specific electrode contact were considered, the mean shift was 0.92 (range: 0.25 to 2.0) electrode contacts, which is comparable to the previous research where shifts of 0.5 to 2.0 (Saoji & Litvak 2010) and 0.08 to 2.01 electrode contacts (Klawitter et al. 2018) were found. The shift in place of excitation was significantly larger at the apex of the electrode array, and also for lateral wall electrode arrays versus (semi) medial wall electrode arrays.

The current study is an addition to the existing literature about phantom stimulation because of its relatively high number of study subjects (15) and thorough testing of the perceived shift in place of excitation following phantom stimulation. The direct comparison with the pitch perceived with steered MP stimulation is advantageous because it enables more accurate quantification of the shift in place of stimulation. Nevertheless, it is extremely difficult for CI users to distinguish pitches and indicate the higher-pitched sound, and this could lead to uncertain results on psychophysical tests concerning pitch, especially at lower stimulus levels. To compensate for this in the current study, only electrode contacts with high pitch discrimination scores were selected for further testing, and an initial pitch matching test was performed to increase the reliability of the shift test in high performing subjects. The downside of this preselection of electrode contacts is that it might induce bias as it could be that higher performing electrodes respond differently to phantom stimulation than poor-performing electrodes. For example, if one assumes that electrodes with good pitch discrimination scores are placed in closer proximity to the auditory nerve, these electrode contacts probably benefit less from phantom stimulation than low scoring electrodes. Nevertheless, a significant correlation was found between pitch discrimination scores and variations in psychometric functions ($R^2 = 0.157$, n = 165, $p < 0.001$). This implies that the selection of high performing electrodes leads to less variation in the pitch shift test, signifying the importance of the pitch discrimination experiment. The pitch matching results were similar to the measured shifts in place of excitation for all SBC configurations at all electrode contacts and for the $PSA_{0.75}$ results taken at the more basally located electrode contacts. Interestingly, the pitch matching and pitch shift measurements did not match for the $PSA_{0.75}$ configuration measured at the more apically located electrode contacts. This could be a consequence of cross-turn stimulation in the apex, as described by Finley et al. (2008) and Frijns et al. (2001), although we cannot explain why this

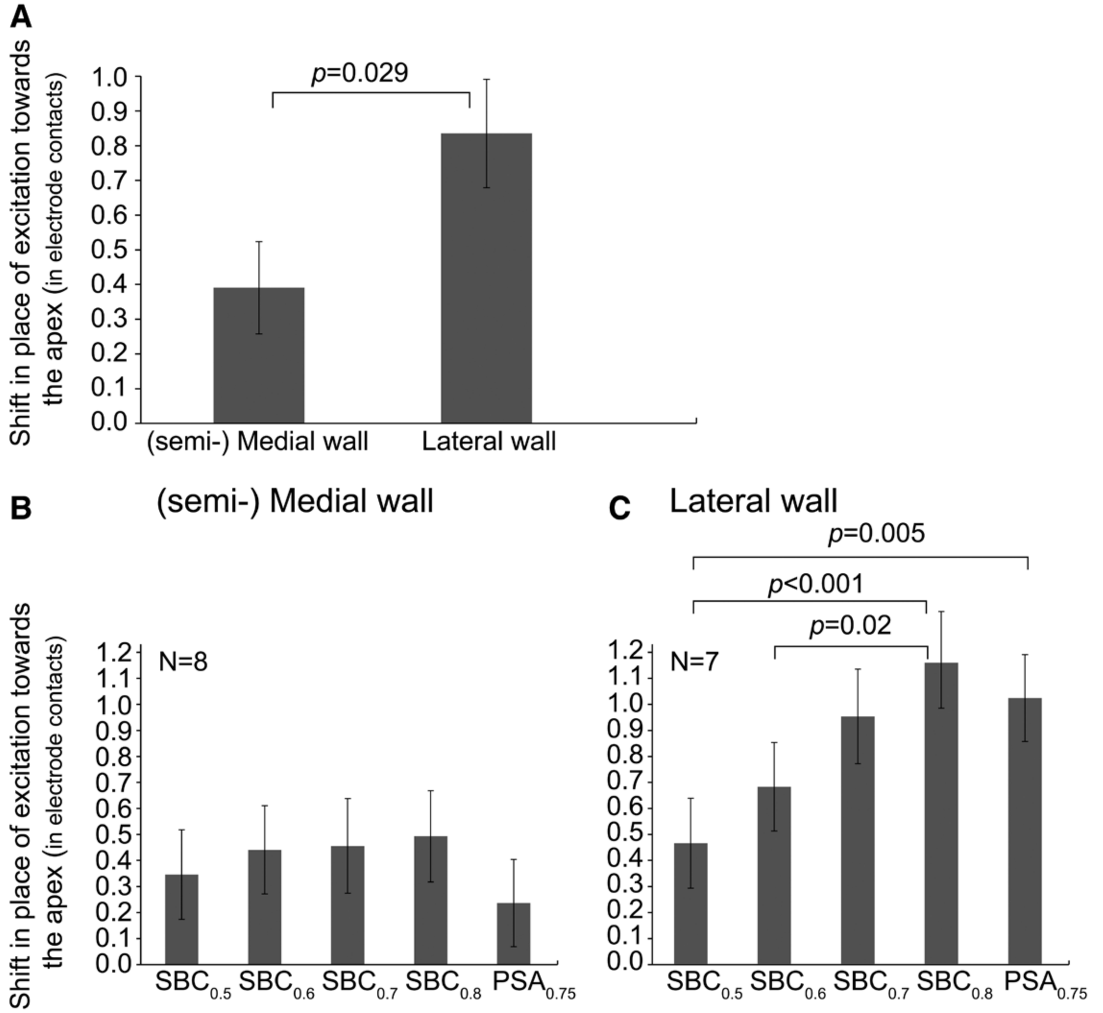

Fig. 4. Effect of electrode type on pitch shift. Mean shift in place of excitation from the main electrode contact for the (semi)medial wall (i.e., mid-scalar or perimodiolar) and lateral wall electrodes at the M-level (A). The effects of the different phantom configurations are displayed for the (semi)medial wall electrodes (B) and lateral wall electrodes (C) separately. All displayed mean and standard errors were calculated using a linear mixed model analysis with either "electrode type" or "configuration" as factors. The error bars represent the standard error. The displayed means were compared in a more extensive linear mixed model, as described in the Results section, and only the statistically significant *p* values (<0.05) are displayed. PSA indicates pseudomonophasic pulse shape, anodic first; SBC, biphasic cathodic first pulse shape.

is specifically the case for $\text{PSA}_{0.75}$ stimulation and not for the other tested stimulation modes. It could also be that the stimuli being matched differ in some other dimension that makes the test harder in one of the two used methods. For example, PSA stimuli may produce narrower excitation patterns than (steered) MP stimuli.

When comparing the shifts in place of excitation with the different phantom configurations, we specifically looked at the apical measurements because a phantom electrode contact was originally intended to be implemented at the most apical location of the electrode array, although the strategy could also be useful at other locations along the electrode array. For example, when arrays are deeply implanted into the cochlea phantom stimulation could be used to stimulate the most basal part of the cochlea that otherwise could not be reached by the most basally located electrode contact. The $\text{SBC}_{0.8}$, $\text{SBC}_{0.7}$, and $\text{PSA}_{0.75}$ configurations showed the largest shifts, with approximately the same average shift of 0.75 electrode contacts toward the apex. Nevertheless, the variation between subjects differed for the three best performing configurations with the $\text{SBC}_{0.8}$ configuration resulting in the most variation between subjects, followed by $\text{PSA}_{0.75}$, while $\text{SBC}_{0.7}$ was the most constant across subjects (Fig. 2). For that reason, one could conclude that $\text{SBC}_{0.7}$ would be the most convenient configuration to implement in a speech coding strategy, in a one-fits-all construction. On the other hand, the greatest shifts in place of excitation were achieved with $\text{PSA}_{0.75}$ (up to two electrode contacts), implying that a larger gain could be achieved for some subjects. Therefore, we recommend an individual fitting, in which a pitch ranking experiment is performed with all three configurations to determine which should be implemented in the final speech coding strategy. The reason for the greater variation between subjects with $\text{SBC}_{0.8}$ and $\text{PSA}_{0.75}$ probably has to do with individual differences in electrode location. The $\text{SBC}_{0.8}$ setting has a relatively high amplitude on the compensating electrode contact. If this contact lies relatively close to the spiral ganglion cells, it could stimulate the auditory nerve on its own, counteracting the electrical field shaping with phantom stimulation in some subjects. The

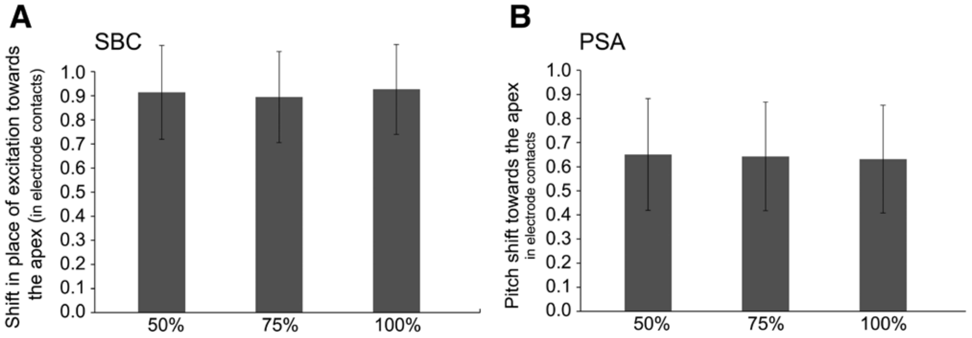

Fig. 5. The effect of loudness on pitch shift. Mean shift in place of excitation from the main electrode contact toward the apex at different loudness levels for the biphasic phantom configuration (A) and the pseudomonophasic configuration (B), both calculated using a linear mixed model with "loudness level" as the main effect. The loudness levels are calculated as percentages of the amplitudes at the most comfortable loudness. For each subject, only the biphasic configurations that had the largest shift in place of excitation at 100% of the most comfortable level was measured at all loudness levels. There was no significant difference in shift between loudness levels. Error bars represent the standard error. PSA indicates pseudomonophasic pulse shape, anodic first; SBC, biphasic cathodic first pulse shape.

$\text{PSA}_{0.75}$ setting has a larger distance between the two involved electrode contacts, which could also influence the effectiveness of electrical field shaping.

Subjects with a lateral wall electrode achieved greater shifts in place of excitation (in electrode units) than those with a (semi) medial wall electrode array, which was also predicted from our computational model of the human cochlea (Snel-Bongers 2013). In (semi) medial wall electrode arrays, the electrode contacts are placed relatively close to the auditory nerve (Van Der Jagt et al. 2016), while a certain distance between the electrodes and the spiral ganglion cells is necessary to effectuate electrical field shaping. This has also been shown in other strategies that make use of electrical field shaping, for example, current focusing that shapes the electrical field to increase spatial selectivity. In a computational model of the human cochlea, both Kalkman et al. (2015) and Goldwyn et al. (2010) demonstrated that closer proximity to the spiral ganglion cells results in less effective current focusing. Moreover, the close proximity can cause neuronal excitation by the compensating electrode already at low degrees of phantom stimulation (Litvak et al., 2007), canceling out the shift toward the apex. This is consistent with the fact that an increase in the degree of phantom stimulation does not increase the shift in place of excitation for the (semi) medial wall electrodes, but that a clear trend is visible for the lateral-wall electrodes (Fig. 4B and C).

Interestingly, the effect of electrode location was also significant, as a larger shift in place of excitation (in electrode contacts) was observed at the apex than at the base. While this is beneficial for the implementation of phantom stimulation in a speech coding strategy, as one would use phantom stimulation specifically on the most apical electrode contact, the reason for it is uncertain. It could be that the electrode contacts at the apex have a larger distance to the auditory nerve than the basal electrodes, although a linear mixed model with the factors "distance to the inner wall" and "electrode number" did not reveal a significant effect (regression coefficient = −0.0021, $p = 0.56$). Another hypothesis that was not addressed in the current study was that the neural survival at the apex is better than at the base of the cochlea (Bierer 2007), resulting in lower thresholds that could decrease the chance for side lobe activation (Kalkman et al. 2015). Nevertheless, it is likely that neural survival plays an important role in the effect of phantom stimulation. For example, if the compensating electrode is in a dead region (Moore 2004), it may not excite nearby auditory nerve fibers; thus, potential side lobes will not cause neural excitation and will have no detrimental effect, even for high σ values. In this study, the data are analyzed based on the electrode shifts. However, contact spacing differs among the three different electrode array types. Furthermore, the angle of insertion depends on depth, modiolar location, contact spacing, and size or geometry of the cochlea. A shift of a certain number of contacts in the apex may not result in the same perceptual difference in the base. For example, Klawitter et al. (2018) analyzed the shift from MP to phantom stimulation in electrode units and octave units and found the two analyses produced very different results. This is not surprising because the magnitude of perceptual differences between contacts varies across the array (and shrinks into the apex) (Landsberger et al. 2014). This could also be an explanation for the different pitch shifts at the apical and the basal part of the electrode array which were found in the current study.

In contrast to previous studies, the effect of phantom stimulation on perceived pitch was similar at different stimulus levels in the current study. The computational model of the human cochlea predicted a decrease in shift in place of excitation at lower stimulus levels. The hypothesis was that the steering of the center of excitation is caused by a suppression of the excitation on the basal side (compensating electrode), while the fibers at the main contact are still excited. At low levels, the number of fibers that can be suppressed on one side is smaller; thus, the phantom effect is diminished. Although this is a plausible hypothesis, the current study did not confirm this, and the effect of stimulus level was not clear from previous reports. Moreover, an alternative hypothesis is that at low overall levels, the basal side-lobe would fall below threshold, leading to larger pitch shifts. Nevertheless, if stimulus level has a limited effect on pitch, indeed, this leads to easier implementation of phantom stimulation in speech coding strategies.

Future Perspectives

The phantom electrode technique results in a shift of the place of excitation, indeed. Previous studies reported positive effects on speech perception when incorporating it in a speech coding strategy (Munjal et al. 2015), although the phantom strategy used by Nogueira et al. (2015) also conveyed lower frequencies, so the beneficial effect might not solely come from the incorporation of a phantom electrode contact. However, to minimize the risk of unwanted excitation near the compensating electrode contact that may cause a pitch reversal or dual-tone, mostly phantom-based speech coding strategies that use relatively low σ values were studied. For example, Sigma's of 0.5 to 0.625 were used in the study of Nogueira et al. (2015). The results of the current study imply that the beneficial effect can be even greater when higher *σ* values are used. Because the best configuration is different for each individual subject and electrode contact, it might be helpful to perform a pitch ranking or pitch discrimination test before fitting subjects with a phantom strategy, like was done in the study of Nogueira et al. (2015). In that study, however, the highest possible sigma value was 0.625, while potentially sigmas up to 1.0 could be beneficial in some subjects. This individualization of speech coding strategies might be advantageous not only for phantom stimulation but also for other speech coding strategies. Moreover, the perceived shift in place of excitation following phantom stimulation is greatest in CI users with a lateral wall electrode array, and this should be considered during clinical implementation.

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M.A.M.d.J. designed the study and carried out the experiments with help from G.R.F.M. Timp under supervision of J.J.B. and J.H.M.F.. J.D.B. created the specific experiments with the use of Matlab, S.B. verified the statistical methods, and J.S.-B., J.J.B., and J.H.M.F. supervised the findings and interpretation of this work. All authors discussed the results and commented on the manuscript.

The authors have no conflicts of interest to disclose.

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REFERENCES

Arnoldner, C., Kaider, A., Hamzavi, J. (2006). The role of intensity upon pitch perception in cochlear implant recipients. *Laryngoscope, 116*, 1760–1765.

Arnoldner, C., Riss, D., Baumgartner, W., et al. (2007). Cochlear implant channel separation and its influence on speech perception – Implications for a new electrode design. *Audiol Neurotol*, *12*, 313–324.

Bierer, J. A. (2007). Threshold and channel interaction in cochlear implant users: evaluation of the tripolar electrode configuration. *J Acoust Soc Am, 121*, 1642–1653.

Biesheuvel, J. D., Briaire, J. J., de Jong, M. A. M., et al. (2019). Channel discrimination along all contacts of the cochlear implant electrode array and its relation to speech perception. *Int J Audiol, 58*, 262–268.

Buchman, C. A., Dillon, M. T., King, E. R., et al. (2014). Influence of cochlear implant insertion depth on performance: A prospective randomized trial. *Otol Neurotol, 35*, 1773–1779.

Büchner, A., Illg, A., Majdani, O., et al. (2017). Investigation of the effect of cochlear implant electrode length on speech comprehension in quiet and noise compared with the results with users of electro-acoustic-stimulation, a retrospective analysis. *PLoS One, 12*, e0174900.

Carlyon, R. P., Lynch, C., Deeks, J. M. (2010). Effect of stimulus level and place of stimulation on temporal pitch perception by cochlear implant users. *J Acoust Soc Am, 127*, 2997–3008.

Carlyon, R. P., Monstrey, J., Deeks, J. M., et al. (2014). Evaluation of a cochlear-implant processing strategy incorporating phantom stimulation and asymmetric pulses. *Int J Audiol, 53*, 871–879.

Cosetti, M. K., & Waltzman, S. B. (2012). Outcomes in cochlear implantation: variables affecting performance in adults and children. *Otolaryngol Clin North Am, 45*, 155–171.

Finley, C. C., Holden, T. A., Holden, L. K., et al. (2008). Role of electrode placement as a contributor to variability in cochlear implant outcomes. *Otol Neurotol, 29*, 920–928.

Firszt, J. B., Koch, D. B., Downing, M., et al. (2007). Current steering creates additional pitch percepts in adult cochlear implant recipients. *Otol Neurotol, 28*, 629–636.

Frijns, J. H., Briaire, J. J., Grote, J. J. (2001). The importance of human cochlear anatomy for the results of modiolus-hugging multichannel cochlear implants. *Otol Neurotol, 22*, 340–349.

Gani, M., Valentini, G., Sigrist, A., et al. (2007). Implications of deep electrode insertion on cochlear implant fitting. *J Assoc Res Otolaryngol, 8*, 69–83.

Goldwyn, J. H., Bierer, S. M., Bierer, J. A. (2010). Modeling the electrode-neuron interface of cochlear implants: Effects of neural survival, electrode placement, and the partial tripolar configuration. *Hear Res, 268*, 93–104.

Harel, O., & Zhou, X. H. (2007). Multiple imputation: Review of theory, implementation and software. *Stat Med, 26*, 3057–3077.

Hochmair, I., Arnold, W., Nopp, P., et al. (2003). Deep electrode insertion in cochlear implants: apical morphology, electrodes and speech perception results. *Acta Otolaryngol, 123*, 612–617.

Kalkman, R. K., Briaire, J. J., Dekker, D. M., et al. (2014). Place pitch versus electrode location in a realistic computational model of the implanted human cochlea. *Hear Res, 315*, 10–24.

Kalkman, R. K., Briaire, J. J., Frijns, J. H. (2015). Current focussing in cochlear implants: an analysis of neural recruitment in a computational model. *Hear Res, 322*, 89–98.

Klawitter, S., Landsberger, D. M., Büchner, A., et al. (2018). Perceptual changes with monopolar and phantom electrode stimulation. *Hear Res, 359*, 64–75.

Kenway, B., Tam, Y., Vanat, et al. (2015). Pitch Discrimination: An Independent Factor in Cochlear Implant Performance Outcomes. *Otol Neurotol, 36*:1472–1479.

Kong, Y. Y., Cruz, R., Jones, J. A., et al. (2004). Music perception with temporal cues in acoustic and electric hearing. *Ear Hear, 25*, 173–185.

Landsberger, D. M., Mertens, G., Punte, A. K., et al. (2014). Perceptual changes in place of stimulation with long cochlear implant electrode arrays. *J Acoust Soc Am, 135*, EL75–EL81.

Landsberger, D. M., Svrakic, M., Roland, J. T. Jr, et al. (2015). The relationship between insertion angles, default frequency allocations, and spiral ganglion place pitch in cochlear implants. *Ear Hear, 36*, e207–e213.

Landsberger, D. M., Vermeire, K., Claes, A., et al. (2016). Qualities of single electrode stimulation as a function of rate and place of stimulation with a cochlear implant. *Ear Hear, 37*, e149–e159.

Litvak, L. M., Spahr, A. J., Emadi, G. (2007). Loudness growth observed under partially tripolar stimulation: Model and data from cochlear implant listeners. *J Acoust Soc Am, 122*, 967–981.

Macherey, O., & Carlyon, R. P. (2012). Place-pitch manipulations with cochlear implants. *J Acoust Soc Am, 131*, 2225–2236.

Macherey, O., Carlyon, R. P., van Wieringen, A., et al. (2008). Higher sensitivity of human auditory nerve fibers to positive electrical currents. *J Assoc Res Otolaryngol, 9*, 241–251.

Macherey, O., Deeks, J. M., Carlyon, R. P. (2011). Extending the limits of place and temporal pitch perception in cochlear implant users. *J Assoc Res Otolaryngol, 12*, 233–251.

Moore, B. C. (2004). Dead regions in the cochlea: conceptual foundations, diagnosis, and clinical applications. *Ear Hear, 25*, 98–116.

Munjal, T., Roy, A. T., Carver, C., et al. (2015). Use of the phantom electrode strategy to improve bass frequency perception for music listening in cochlear implant users. *Cochlear Implants Int, 16* (Suppl 3), S121–S128.

Nogueira, W., Litvak, L. M., Saoji, A. A., et al. (2015). Design and evaluation of a cochlear implant strategy based on a "Phantom" channel. *PLoS One, 10*, e0120148.

Pijl, S. (1997). Pulse rate matching by cochlear implant patients: effects of loudness randomization and electrode position. *Ear Hear, 18*, 316–325.

Potts, L. G., Skinner, M. W., Gotter, B. D., et al. (2007). Relation between neural response telemetry thresholds, T- and C-levels, and loudness judgments in 12 adult nucleus 24 cochlear implant recipients. *Ear Hear, 28*, 495–511.

Qi, B., Liu, B., Krenmayr, A., et al. (2011). The contribution of apical stimulation to Mandarin speech perception in users of the MED-EL COMBI 40+ cochlear implant. *Acta Otolaryngol, 131*, 52–58.

Rak, K., Schraven, S. P., Schendzielorz, P., et al. (2017). Stable longitudinal performance of adult cochlear implant users for more than 10 years. *Otol Neurotol, 38*, e315–e319.

Saoji, A. A., Landsberger, D. M., Padilla, M., et al. (2013). Masking patterns for monopolar and phantom electrode stimulation in cochlear implants. *Hear Res, 298*, 109–116.

Saoji, A. A., & Litvak, L. M. (2010). Use of "phantom electrode" technique to extend the range of pitches available through a cochlear implant. *Ear Hear, 31*, 693–701.

Snel-Bongers, J. (2013). *Dual electrode stimulation in cochlear implants: From concept to clinical application (Doctoral thesis)*. Leiden University, The Netherlands.

Snel-Bongers, J., Briaire, J. J., Vanpoucke, F. J., et al. (2012). Spread of excitation and channel interaction in single- and dual-electrode cochlear implant stimulation. *Ear Hear, 33*, 367–376.

Snel-Bongers, J., Briaire, J. J., van der Veen, E. H., et al. (2013). Threshold levels of dual electrode stimulation in cochlear implants. *J Assoc Res Otolaryngol, 14*, 781–790.

Townshend, N., Van Compernolle, D., White, R.L. (1987). Pitch perception by cochlear implant subjects. *J Acoust Soc Am*, *82*, 106–115.

van der Beek, F. B., Boermans, P. P., Verbist, B. M., et al. (2005). Clinical evaluation of the Clarion CII HiFocus 1 with and without positioner. *Ear Hear, 26*, 577–592.

Van Buuren, S. (2018). *Flexible Imputation of Missing Data*, Second Edition, Chapman and Hall/CRC.

van der Jagt, M. A., Briaire, J. J., Verbist, B. M., et al. (2016). Comparison of the HiFocus Mid-Scala and HiFocus 1J electrode array: Angular insertion depths and speech perception outcomes. *Audiol Neurootol, 21*, 316–325.

van der Marel, K. S., Briaire, J. J., Wolterbeek, R., et al. (2014). Diversity in cochlear morphology and its influence on cochlear implant electrode position. *Ear Hear, 35*, e9–20.

von Ilberg, C. A., Baumann, U., Kiefer, J., et al. (2011). Electric-acoustic stimulation of the auditory system: A review of the first decade. *Audiol Neurootol, 16 Suppl 2*, 1–30.

Wardrop, P., Whinney, D., Rebscher, S. J., et al. (2005). A temporal bone study of insertion trauma and intracochlear position of cochlear implant electrodes. II: Comparison of Spiral Clarion and HiFocus II electrodes. *Hear Res, 203*, 68–79.

Wichmann, F. A., & Hill, N. J. (2001a). The psychometric function: II. Bootstrap-based confidence intervals and sampling. *Percept Psychophys, 63*, 1314–1329.

Wichmann, F. A., & Hill, N. J. (2001b). The psychometric function: I. Fitting, sampling, and goodness of fit. *Percept Psychophys, 63*, 1293–1313.